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A POROUS MEDIA APPROACH FOR FOOT BIOMECHANICS

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Abstract. Recently a new computational model, based on the thermodynamically constrained averaging theory (TCAT), has been proposed to predict tumor initiation and proliferation [1, 2]. A similar mathematical approach is being developed for diabetic ulcer prevention. The common aspects at continuum level are the macroscopic balance equations governing the flow of the fluid phase, diffusion of chemicals and tissue mechanics, and some of the constitutive equations. We show here as first step a porous media model of the foot where the ulcer model will be introduced in future.

TCAT [3] is a framework recently established for the analysis of multiphase systems, which is consistent over multiple scales. It provides a rigorous yet flexible method for developing multiphase, continuum models at any scale of interest. TCAT uses averaging theorems to formally and consistently convert micro-scale equations to the larger macro-scale.

The soft plantar tissue is modeled as a multiphase system consisting of two phases: one solid for the tissue cells and their extracellular matrix (i.e. ECM components and the tissue cells are treated as a single solid phase), and one fluid (the interstitial fluid). The load history and boundary conditions are consistent with experimental measurements performed at the mega level: in-vivo foot kinematics and magnetic resonance give the input data of the model [4]. The governing equations are discretized in space by the finite element method [5] and in time domain by using the θ-Wilson Method. An example of application is presented at the end of the paper where attention is focused on quantities which will be of importance in diabetic ulcer modeling.

1 INTRODUCTION

The infection of foot ulcers in diabetic patients leads often to the limb amputation. It is clear that the prevention of ulcers is the major method of preventing amputations. One of the main goals in diabetic treatments is therefore trying to reduce excessive pressures in the

plantar tissue, since ulceration is initiated by abnormal mechanical loading of the foot. These abnormal load conditions in some areas of the plantar tissue are the consequences of changes in foot structure and in tissues properties connected with diabetic peripheral neuropathy.

To prevent ulcers, the main technique is to off-load the sites that tend to ulcerate (first, second, or third metatarsal head, and less frequently the hell) in the patients where an excessive pressure peak is measured. However, ulceration often starts at the interface between bone angularity and the plantar tissue, probably due to internal stresses, which are not directly evaluable using experimental tools. In this framework, finite element (FE) simulations are of crucial importance to study the pressure peaks in the plantar tissue, because they give the stress field in the whole discretized domain.

Great effort has been made by several research teams to reproduce foot kinematics during gait to obtain numerically the normal pressure measured at the ground surface by experimental devices. However, the plantar tissue, which is directly concerned with ulceration, is almost unanimously modeled as a hyperelastic material [6-11] while it behaves like a viscoelastic material as observed experimentally by Gefen *et al.* [12] and as appropriately assumed by Pai and Ledoux [13] in their model. If the viscous nature of the soft foot tissue is taken into account it is possible to evaluate the influence of time (i.e. the duration of the gait cycle) on tissue strain. This is not possible by using only linear elastic or hyperelastic models.

In this paper a porous media approach to model numerically the gait cycle and the associated plantar pressures is presented. This method can integrate experimental data of the patient-specific foot kinematics into the modeling process, thus improving the agreement between the computed pressure and the measured one. The mathematical model is based on the Thermodynamically Consistent Averaging Theory (TCAT) which is a two-scale procedure. With respect to the existing purely macroscopic formulations, it allows to incorporate information coming from the micro-structure of the plantar tissue, which is modeled as a porous medium saturated by an interstitial fluid. The presence of the interstitial fluid has several implications that, together with the other features of the presented model, are discussed in the next sections.

2 SYNOPSIS OF TCAT

The mathematical model is based on TCAT, which is a rigorous method for developing continuum multiphase models at any scale of interest. An important feature of the procedure is that it explicitly defines larger scale variables in terms of smaller scale variables. When modeling transport in multiphase systems, the length scale of the model impacts the form and parameterization of the governing equations. At the microscale - smallest scale at which the continuum hypothesis holds - a single (continuum) point contains a large number of molecules so that properties such as density, temperature, and pressure of a phase can all be defined. At the microscale, classical “point” conservation equations and thermodynamic expressions are written. However, the domains of many problems of interest are too large, and the phase distributions are too complex to be modeled at the microscale only. The level of detail required to account for geometric structure and the variability of variables at the microscale allows simulation of only very small domains. To overcome this challenge, many porous media models are formulated at a larger scale, called the macroscale, adequate for

describing system behaviour while filtering out the high frequency spatial variability.

The macroscale depends on the concept of the representative elementary volume (REV), an averaging volume that can be centred at each point in the system and which is large enough to include all phases present so that averages are independent of the REV size. The volume must also be sufficiently small so that quantities such as gradients are meaningful. TCAT consistently transforms microscale conservation and thermodynamic equations to the macroscale and converts averages of microscale derivatives into derivatives of macroscale average quantities. The description of a multiphase system must include dynamic conservation and thermodynamic equations for all phases, interfaces (where two phases meet), common curves (where three interfaces meet), and common points (where four common curves meet). Averaging theorems transform equations describing processes in these entities from the microscale to the macroscale (see Gray *et al.*, [14]).

To close the conservation equations - which contain additional terms due to averaging - new model parameters and constitutive relations must be specified. TCAT employs averaged thermodynamic relations to guide closure of the system of equations.

The benefits of using a TCAT approach are as follows. First, the model derivation proceeds systematically from known microscale relations to mathematically and physically consistent larger scale relations. This is accomplished by use of averaging theorems. Second, the thermodynamic analysis is consistent between scales, in the definitions of variables at different scales, and in satisfying the entropy inequality. The interscale consistency and explicit definition of variables are not achieved using a rational thermodynamic approach. Third, relations may be obtained for the evolution of the spaces occupied by phases and of the interfacial area density. These relations are based on the averaging theorems. TCAT is now successfully used for soft tissue modeling, see [1, 2].

3 GAIT CYCLE AND MODELING ASPECTS

As already mentioned the model is at its early stage of development and some of the constitutive relationships governing ulceration must be better investigated. As a consequence, even if the model is developed for diabetic foot, in the example a healthy foot is modeled. When the plantar tissue and its microvasculature are healthy, and the structure and shape of the foot are normal, the foot plays efficiently one of its principal functions, i.e., shock-absorbing capability and adaptation to the ground surface during gait. The gait cycle consists of two basic components: the stance phase, when the foot is in contact with the ground, and the swing phase, when the foot is in the air for limb advancement. The stance (also called weight bearing phase) can be subdivided into three sub-phases: the contact phase, the midstance phase and the propulsion phase [15]. To mimic the evolution of the pressure in the foot by finite element analysis, the gait cycle is here modeled from the end of the contact phase to the active propulsion phase. The global forces measured experimentally by the force platform device, translated and applied with opposite sign to the ankle, are the input load for the numerical model. Applying a patient-specific load history is essential because not only the foot morphology and tissue properties change due to diabetic diseases, but gait alterations may also occur. For example, it has been observed that commonly patients with diabetes walk slower, tend to take shorter steps with a wider base of support, and demonstrate a longer double support time [16].

A faithful representation of the kinematics of bone structures is important to generate the correct stress field in the plantar tissue. To this aim, when a real case is modeled, the first comparisons between numerical simulations and experimental results is performed on foot kinematics during the gait cycle.

As mentioned, the plantar tissue is modeled as a porous medium. In fact it is composed of a lower microchamber layer and an inner macrochamber layer, as it can be easily seen ultrasonographically [17]. The use of a porous medium model has already been exploited in biomechanics simulations, for example to simulate intervertebral discs [18, 19]. In the case of soft plantar tissue, it allows for a more realistic structural modelling, since consolidation phenomena occurring inside the tissue can be taken into account, thus resulting in a global visco-elastic behaviour. Furthermore, effective stress field is changing during consolidation, transferring the load bearing action from the interstitial fluid to the solid skeleton.

4 THE MULTIPHASE MODEL OF THE FOOT TISSUE

The foot tissue is modeled as an elastic porous medium in large strain regime completely filled by a fluid phase. The tissue cells and their extracellular matrix form the solid skeleton *s* with pores saturated by the interstitial fluid *f*. Indeed, the sum of the volume fractions for the two phases has to be unit

$$\varepsilon^s + \varepsilon^f = 1 \tag{1}$$

Being the pores fully saturated, the volume fraction of the interstitial fluid, ε^f, is equal to the porosity of the medium, which is denoted here as ε. Transport of nutrients and possible drugs delivery within the microvasculature can be also considered by means of the introduction of an effective diffusion coefficient which is estimated from the real degree of vascularization of the zone of interest and taking into account the strain field.

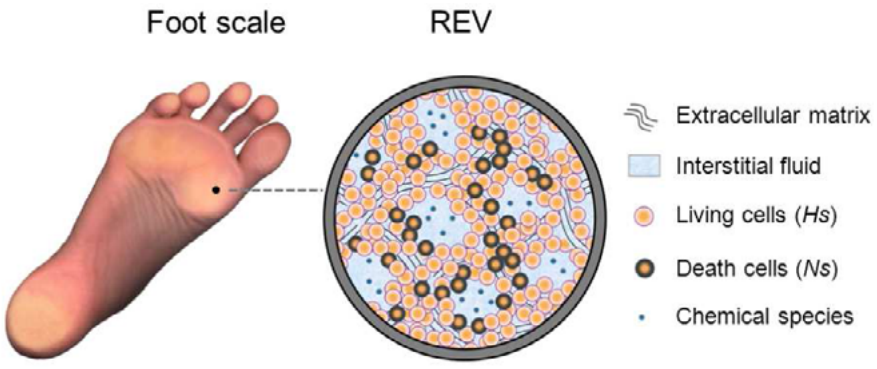

Figure 1: The foot tissue multiphase system: REV and its connection with the foot scale

4.1 The governing equations

The plantar tissue may become necrotic depending on the stress level and on vasculopathy; thus it comprises a healthy fraction (*Hs*) and a necrotic one (*Ns*). The mass conservation equations of the healthy and necrotic fractions of the tissue read respectively

$$\frac{\partial\left(\varepsilon^s\rho^s\omega^{\overline{Hs}}\right)}{\partial t}+\nabla\cdot\left(\varepsilon^s\rho^s\omega^{\overline{Hs}}\mathbf{v}^{\bar{s}}\right)+\varepsilon^s r^{Ns}=0 \tag{2}$$

$$\frac{\partial\left(\varepsilon^s\rho^s\omega^{\overline{Ns}}\right)}{\partial t}+\nabla\cdot\left(\varepsilon^s\rho^s\omega^{\overline{Ns}}\mathbf{v}^{\bar{s}}\right)-\varepsilon^s r^{Ns}+\overset{Ns\to f}{M}=0 \tag{3}$$

where $\omega^{\overline{Hs}}$ is the mass fraction of the healthy tissue cells (and associated ECM), $\omega^{\overline{Ns}}$ is the mass fraction of the necrotic tissue cells (and associated ECM), ρ^s is the density of the tissue and $\mathbf{v}^{\bar{s}}$ is the velocity of the solid phase. The term $\varepsilon^s r^{Ns}$ is the cells' death rate and represents an intra-phase exchange of mass (i.e. within the phase *s*). $\overset{Ns\to f}{M}$ is the rate of dissolution of the necrotic cells and is an inter-phase exchange of mass (from the phase *s* to the phase *f*). Summing equations (2) and (3) gives the mass balance equation of the solid phase

$$\frac{\partial\left(\varepsilon^s\rho^s\right)}{\partial t}+\nabla\cdot\left(\varepsilon^s\rho^s\mathbf{v}^{\bar{s}}\right)+\overset{Ns\to f}{M}=0 \tag{4}$$

From TCAT, the mass conservation equation of the chemical species *i* in the interstitial fluid *f*, reads

$$\frac{\partial\left(\varepsilon^f\rho^f\omega^{\overline{if}}\right)}{\partial t}+\nabla\cdot\left(\varepsilon^f\rho^f\omega^{\overline{if}}\mathbf{v}^{\bar{f}}\right)+\nabla\cdot\left(\varepsilon^f\rho^f\omega^{\overline{if}}\mathbf{u}^{\overline{\overline{if}}}\right)-\overset{is\to if}{M}=0 \tag{5}$$

where $\omega^{\overline{if}}$ is the mass fraction of the species *i* dispersed within the phase *f* (e.g. oxygen, glucose, drugs…), $\overset{is\to if}{M}$ is an inter-phase exchange term (mass of the chemical species *i* consumed or relaxed by the tissue) and $\mathbf{u}^{\overline{\overline{if}}}$ is the diffusive velocity of the species *i*. Summing equation (5) over all species gives the mass balance equation of the interstitial fluid

$$\frac{\partial\left(\varepsilon^f\rho^f\right)}{\partial t}+\nabla\cdot\left(\varepsilon^f\rho^f\mathbf{v}^{\bar{f}}\right)-\overset{Ns\to f}{M}=0 \tag{6}$$

It is assumed that the mass of chemical substances consumed by the cells is equal to that produced due to their metabolism; cells may however dissolve due to necrosis. Hence the source term in equation (6) reads

$$\overset{Ns\to f}{M}=\begin{cases}\sum_{i\in f}\overset{is\to if}{M}=0 & \text{in a healthy tissue}\\ \sum_{i\in f}\overset{is\to if}{M}>0 & \text{when necroisis occurs}\end{cases} \tag{7}$$

Assuming $\rho^f=\rho^s$ after some transformation (see [20] for the full derivation) we obtain

$$\left(\frac{\varepsilon^f}{K_F}+\frac{1-\varepsilon^f}{K_S}\right)\frac{\partial p^f}{\partial t}=\nabla\cdot\left[\frac{\mathbf{k}}{\mu^f}\nabla p^f\right]-\mathbf{1}:\mathbf{d}^{\bar{\bar{s}}} \tag{8}$$

where $1/K_S$ and $1/K_F$ are the solid and liquid phases compressibility respectively, **k** is the intrinsic permeability tensor, μ^f is the dynamic viscosity, p^f is the interstitial fluid pressure (IFP) and $\mathbf{d}^{\bar{\bar{s}}}$ is rate of strain tensor. Equation (8) is one of the governing equations of the model. The second governing equation of the model is the linear momentum balance of the solid phase in rate form

$$\nabla \cdot \left(\frac{\partial \mathbf{t}^{\bar{\bar{s}}}_{eff}}{\partial t} - \frac{\partial \left(\alpha p^f \right)}{\partial t} \mathbf{1} \right) = 0 \tag{9}$$

where α is the Biot's coefficient and $\mathbf{t}^{\bar{\bar{s}}}_{eff}$ is the effective stress in the sense of porous media mechanics. In this first version of the mathematical model diffusion of oxygen and drug delivery *via* the micro-vasculature are not considered.

4.2 Numerical solution

The chosen primary variables of the model are the interstitial fluid pressure (IFP), p^f, and the displacement vector of the solid phase, $\mathbf{u}^s$. The standard Galerkin procedure is used to discretize in space equations (8) and (9) [5], while for the time discretization the Crack-Nicolson method is used. Thence, the primary variables are expressed in terms of their nodal values as

$$p^f(t) \cong \mathbf{N}_f \bar{\mathbf{p}}^f(t) \qquad \mathbf{u}^s(t) \cong \mathbf{N}_u \bar{\mathbf{u}}^s(t) \tag{10}$$

where $\bar{\mathbf{p}}^f(t)$ and $\bar{\mathbf{u}}^s(t)$ are vectors of nodal values of the primary variables at the time instant t, and N_f, and N_u are vectors of shape functions related to these variables. After the FE discretization the final system of equations can be expressed in a matrix form as follows

$$\mathbf{C}_{ij}(\mathbf{x}) \frac{\partial \mathbf{x}}{\partial t} + \mathbf{K}_{ij}(\mathbf{x}) \mathbf{x} = \mathbf{f}_i(\mathbf{x}) \tag{11}$$

where $\mathbf{x}^{\mathrm{T}} = \left\{ \bar{\mathbf{p}}^f, \bar{\mathbf{u}}^s \right\}$, $\mathbf{C}_{ij}(\mathbf{x})$ and $\mathbf{K}_{ij}(\mathbf{x})$ are the capacity and conduction matrices respectively.

For the solution of the system of equation (11) a staggered scheme is adopted. The convergence properties of such staggered schemes have been investigated by Turska *et al*. [21]. In particular, for the iteration convergence within each computational step a lower limit of $\Delta t/h^2$, function of the material properties, has to be observed; Δt is the time step and h the element size (for detail on the numerical solution see [20]). The model has been implemented in the code Cast3M (*http://www-cast3m.cea.fr*) of the French Atomic Energy Commission.

5 A NUMERICAL EXAMPLE

In this example a healthy foot during a gait cycle is modeled. This is not a real experimental case but a numerical example to show the capability of the model and the interesting aspects which can be captured by modeling the plantar tissue as a porous medium. In future the geometry of a patient's foot (obtained from MR-image) and patient-specific

parameters evaluated experimentally from the properties of the soft tissue and of its micro-vascular network (to take into account the nutrient diffusion through the tissue) will be used.

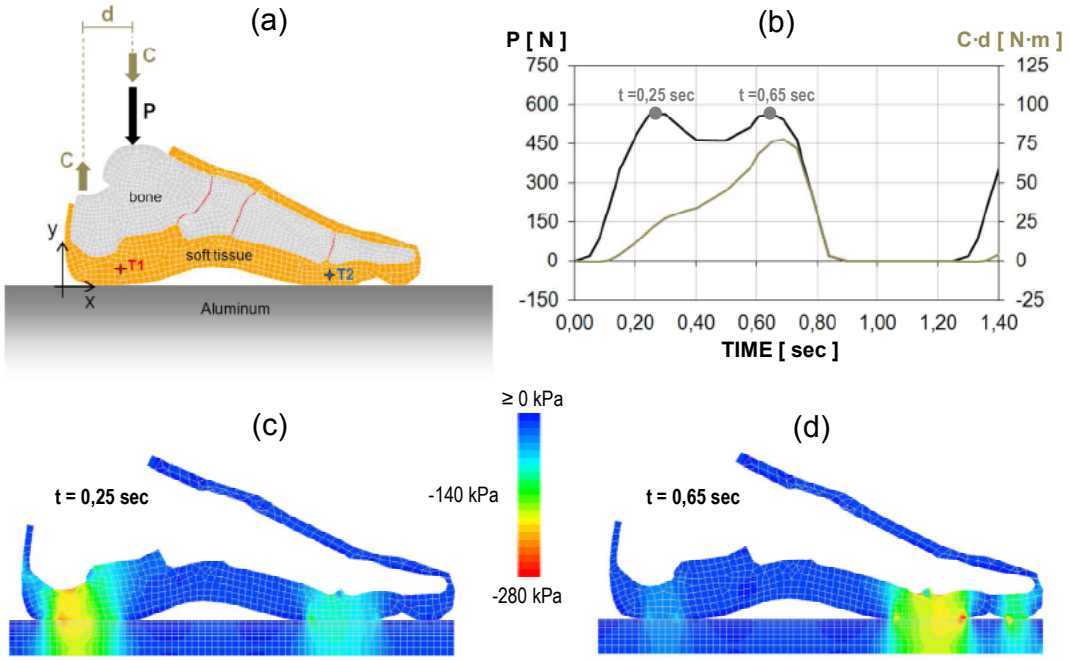

Figure 2: Geometry and load conditions (a); typical load history used in the numerical simulation (Natali *et al.* [11]) (b). Total stress field t_{yy} at 0.25 sec. (c) Total stress field t_{yy} at 0.65 sec.(d).

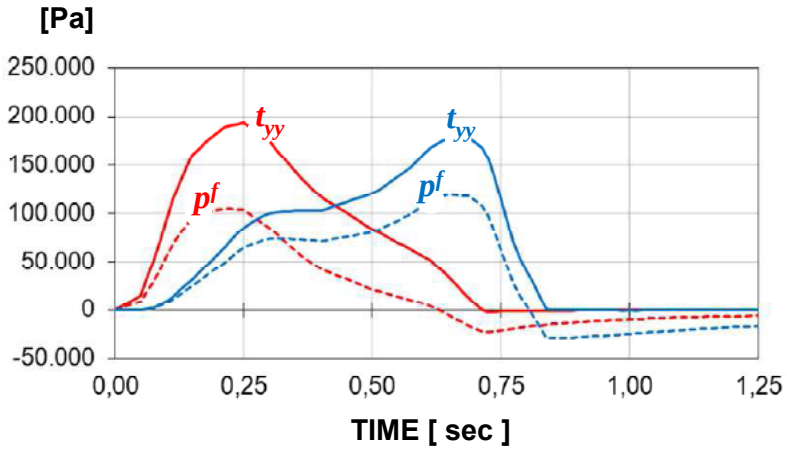

Figure 3: Evolution with time of the total stress t_{yy} and of the interstitial fluid pressure p^f in the points T1 and T2 in figure 2.a.

The finite element mesh of the soft tissue (in orange), of the bone segments (in gray) and of the cartilages (in red) are represented in figure 2.a; an aluminum surface supporting the foot is also modeled. The elastic modulus is selected as E = 1 MPa. The assumed Young' modulus is higher than values measured in some slow-loading plantar tissue studies [12]: it

compensates for neglecting high strain rate stiffening effects and accounts for the increase of the elastic modulus at large strains in hyperelastic tissue behavior, expected to occur in the plantar tissue during gait (similar values are used in [22, 23]). No experimental data have been found in literature for the intrinsic permeability, k, of the plantar tissue. In this work, the intrinsic permeability is assumed as $k = 2 \cdot 10^{-13}$ m^2, the same order of magnitude found in [24], where a similar tissue is analyzed. The other parameters are taken from literature [6-11]. Figure 2.b shows the evolution with time of the force P and of the moment $C \cdot d$. This last one can be computed from the position of the plantar pressure resultant at different times and is applied to the foot as shown in figure 2.a.

Figures 2.c and 2.d show the total vertical stress at 0.25 seconds and at 0.65 seconds respectively. In the first case the loaded area is the heel while in the second case is that of the forefoot; in both cases the peak of pressure is about -280 kPa. The evolution with time of the total stress t_{yy} and of the interstitial fluid pressure p^f in the points T1 (red lines) and T2 (blue lines) are depicted in figure 3 (points T1 and T2 are represented in figure 2.a). The difference between the continuous line and the dashed one is the effective vertical pressure.

6 CONCLUSIONS

A two-phase model for the analysis of the mechanical behavior of the plantar tissue during gait is presented. The mathematical formulation of the model is based on TCAT, a thermodynamically consistent upscaling procedure. An application shows the potential of the method and the resulting fields made available by the numerical tool. The presence of the interstitial fluid in the pores allows to mimic the real global viscoelastic behavior of the plantar soft tissue. This model is at its early stage of development, therefore the numerical results here presented are preliminary. Nevertheless, numerical simulations give values of the pressure peaks and the distribution of stresses on the foot plantar tissue which are in good agreement with those found in literature. Furthermore, even if porous media mechanics is not a novelty in biomechanical simulations, so far to our best knowledge it has never been applied to plantar tissue modeling and prediction of diabetic wounds.

It is worth to point out that input data for this simulation are taken form literature; indeed in a real case the geometry of the patient's foot (obtained from MR-image) and patient-specific parameters must be used to obtain predictive results. Moreover, the in-vivo measurements of displacements and rotations of the bone segments can be useful to check that the kinematics of the foot obtained numerically is the sound one. In a future enhancement, the impact of strain on tissue permeability will be considered. It may have a non negligible impact on the solution in large strain regime. Finally, the tissue vascularization and its efficiency will be taken into account, since the main purpose of this work is to have a numerical tool which can be helpful to better understand the bio-chemo-mechanical processes involved in foot ulceration and its progression.

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